DESIGN DEVELOPMENTS OF SURGICAL SIMULATORS-PASSIVE/ACTIVE AND COUPLED/DECOUPLED AXES DESIGNS

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Abstract - **Advancements in surgical simulators, especially in angioplasty simulation process, determination of necessary parameters and new comprehensive designs for the operation will be reported. First, design considerations and parameters affecting haptic perception, especially for angioplasty operation, will be discussed. After the review of existing simulators, passively actuated haptic interface, HUI, will be presented for its capabilities. Passive actuation of the tools results in an uncompensatable low transparency for the device. Mentioning the drawbacks of passive actuation, a new conceptual design, which enables decoupling of driven axes and active-actuation, will be presented.**

Keywords- **Angioplasty Simulation, Robotic Manipulators, Surgery**

I. INTRODUCTION

Haptics is a sub-area of robotics, which deals with reflected forces to a human operator by virtual- or tele-environment. To enable haptic interaction with the environment, a device called haptic interface is needed which can represent some haptically sensed characteristics of objects in the medium such as shape, stiffness, surface friction and texture, without significantly impeding hand motion, [1],[2]. Merging haptic interfaces with virtual reality has introduced variety of application possibilities that enables training of astronauts, medical doctors, etc.

About the studies conducted over surgical simulators, which is essential for training of doctors, so far, researchers mostly have focused on MIS and rarely angioplasty simulation. Actual surgery contains delicate maneuvers, excellent hand-eye coordination, and above-average hand dexterity and positional accuracy, [3].

A. Angioplasty Simulation and Simulator Needs

CAD (Coronary Artery Disease) is caused by the obstruction of heart arteries by plaque, which is mixed substance, composed of debris and living cells. Current treatments for this disease include interventions to restore blood flow, such as angioplasty (PTCA - Percutinius Transluminal Coronary Angioplasty - removal or compression of the plaque) and CABG - Coronary Artery Bypass Graft Surgery (detouring around the blockages).

Since its first application in September 1977 by Andreas Gruentzig, usage of angioplasty has grown rapidly. Actual procedure begins with an incision in upper inner thigh or groin region. First tube to be inserted is sheath which serves as an initial path to the wire and catheter. Next, the catheter is slided into the sheath through the artery and guided up to the vascular tree to the site of interest. During the operation, the interventionist relies on his/her haptic perception and the fluoroscope visual images (real time radiological imaging), [4]. Coronary revascularization is an expensive method and is applied very frequently, 666,000 angioplasty operations and 598,000 bypass surgeries in 1996, [5]. Also, in 1993, ACC (American College of Cardiology) and AHA (American Heart Association) approved the number of annual interventions to be performed by an interventionist as 75 cases to keep the competency [6].

These studies strongly support that more surgeons should be trained to apply angioplasty to meet the increasing need for the operation.

B. Existing Angioplasty Simulators

In Angioplasty Simulator, [6], simulation of comprehensive angioplasty procedure is aimed. In order to simulate the complete procedure, it has three stages representing wire, catheter and sheath simulation of the actual angioplasty intervention. While catheter and wire parts enable two degree of freedom force activation, sheath part only has unidirectional uncontrolled force reflection and axial position tracking. Hence, the force-reflecting device totally has 5 degree of freedom (DOF). With these features, the device includes all necessary force feedback necessary for the simulation mechanically. However, there was too much force generation and not enough resolution.

The complete endoscopic surgery simulation system, Virtual Endoscope System (VES), [7], has been developed by a group in Nagoya University, Department of Micro System Engineering. Originally, despite the fact that the mechanism is for endoscopic simulation, it is applicable to angioplasty simulation. The device employs four rollers holding the endoscope, two for translational drive and two for rotational drive, meaning 2 DOF system. Each pair of rollers is driven by a motor through a differential mechanism, which reduces the complexity of the control concept.

C. Detailed Design Considerations of an Angioplasty Simulator

Primary concern when designing a haptic interface is human perception capabilities for both motion and force, so that virtual objects can be felt with enough resolution and transparency. With resolution, it is meant that the device's reflected F_{JND}(Force Just Noticeable Difference) is under human force perception resolution. Additionally, transparency refers to the exact force

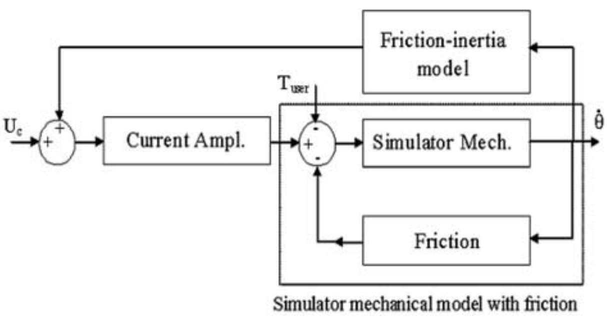

Fig. 1 Friction compensation using velocity feedback

reflection from virtual or tele-environment. Then, based on these limitations, the device can be designed and mimic desired motions and forces, which are generated during actual operations. Human hand dexterity has been addressed by several researchers, [8],[9]. Psychophysical studies shows that sensitivity of human haptic perception is around micron level with the bandwidth of several hundred Hz. This requires a device with high stiffness, low inertia, powerful and responsive actuators, high-resolution sensors and a control system with low time delay, [10].

In all mechanical designs, one way to consume less energy is to reduce the total mass of the system, so that power consumption will be minimal. Inertia causes additional force to user's hand and for instance, when moving in virtual space, user can not observe free space completely. Consequently, inertia should be minimized or using active drive system, it should be compensated.

However, threshold sensitivity and resolution of human hand when sensing reaction forces will allow us to leave some inertia in the design. This provides a designer some safety margin to utilize from. In several haptic interfaces designed up to this time, lowering the inertia below some level is generally aimed. For instance, The force felt by the user throughout the working space is held under 100 gr. in the design of PHANToM, [11]. Also, it has been reported that human force tracking resolution ranges from 11% to 15% for given target forces. These values are improved with visual feedback. Worst case in force tracking is when user follows sinusoidal forces. This is an expected result. However, what is surprising for this is that amount of the error does not vary with target force change, [12].

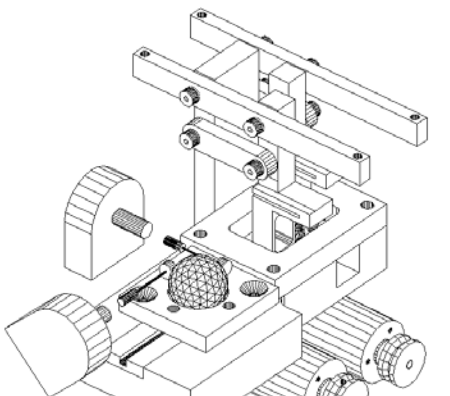

Fig. 2. Modified HUI mechanical system that shows force and position sensor locations

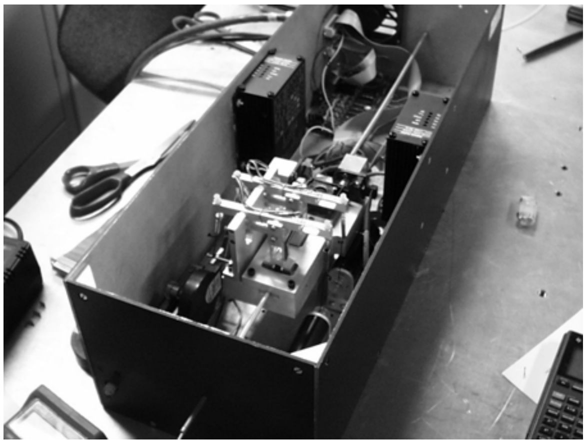

Fig. 3. HUI electromechanical actuation

Likewise, F_{JND} is provided in literature for human force sensing as 7% regardless of test conditions, body sites and reference forces, [13]. Based on these findings, effective inertia and friction in the system should be kept under 5% of the max force exertable by the interface.

Another type of force generation caused by imperfectness of the device is local friction, which originates extra force on user's hands. Especially in geared systems, friction significantly affects the torque applied to the system. Disturbance effects because of the friction in the system can be canceled out online, Fig. 1.

In Fig. 4, a trajectory of a given PTCA simulator is given with certain assumptions, employing friction compensation. Here, a- torque trajectory is given in terms of both position and velocity of the simulator, b- velocity of the catheter is assumed constant in reverse and forward directions for simplicity of the representation, c- there exists one major turn on the way to the coronary artery blockage and d- torque values are assumed constant all the way.

PTCA simulator should reflect any level of stiffness, which occurs during real interventions, to the user's hands. On the other hand, the stiffness to convince the user that an object is rigid should be clearly defined. Based on the tests on the several subjects, the range of 153 to 415 Newton/cm stiffness is provided,[13], for the required stiffness of haptic interfaces.

Overall performance of a haptic interface is described by its bandwidth which is the measure of the highest frequency of the information that can travel across the system. A component (actuator, sensor, controller or amplifier) which has the lowest bandwidth in the system determines the bandwidth of the device,[14].

The difference between minimum ineliminable friction force and the maximum force exertable by the mechanism-actuator system states the force dynamic range. Some researchers set friction threshold to 5% of maximum exertable force as a design goal, [15].

II. MODIFIED HAPTIC USER INTERFACE (HUI)

The development of an angioplasty simulator using passive actuation methods has been accomplished with

HUI design, Fig. 3. This design is intended to simulate only guiding catheter part of the angioplasty operation.

The foremost developments aimed in this design are to improve force generation part, increasing mechanical stability using 4-bar linkage system and, from control system point of view, increasing stability of servo system.

One of the project goals in HUI was to decouple translational and rotational axis keeping passive actuation, so that both axes can be independently controlled. In HUI, catheter moves axially between two rollers in order to create rotational resistance. Likewise, translational part does not resist to rotational movement because of the plunger's saw type cross-section.

As can be viewed from Fig. 2, actuation mechanism of the HUI consists of two 4-bar linkages powered by DC motors and supported by main and auxiliary springs for rigidity considerations and to avoid force components other than vertical.

The use of three springs is needed in order to eliminate existing disturbances in the system and to increase the speed of response of the simulator. Additionally, to eliminate frictional forces between plungers and the catheter when no force is applied, second auxiliary spring can be used to remove contact of the plunger, so that friction created by the weight of the 4-bar linkage will be avoided, Fig. 5.

2000 line optical encoders integrated with the actuators are used to measure the rotational position of the DC motor that actuates the springs and resulting spring position actuation accuracy is 0.022 mm (0.000866”). Based on the calibration curves, applicable force accuracy in axial direction is ±0.375N (0.084 lbf) and applicable torque accuracy around rotational axis is ±0.075 N-cm (0.0066 lbf-in).

Two-axis position measurement system is employed utilizing a mouse ball as shown in Fig. 2. 1024 line encoder for translational axis and 500 line encoder for rotational axis with quadrature coding are used for this purpose. Resulting motion measurement resolutions are 0.196 degrees for rotational axis and 4.93 μm for translational axis.

Force measurement is implemented in original HUI for informative purposes to determine the user forces by placing two sensors on the plungers, Fig. 2 and Fig. 3.

The main difference between HUI and angioplasty simulator that is mentioned in haptic interfaces background is in actuation part, in which high precision Maxon brushless dc motors and servo amplifiers and a Galil 4 axis programmable controller are employed. Having these components separate instead of compact in a motor eliminates possible bugs in the code and provides the power of establishing desired settings independently.

III. PROPOSED IDEAL DESIGN FOR AN ANGIOPLASTY SIMULATOR

Since, in angioplasty operation, there exist two primary axes for each tool, guiding catheter, sheath or wire, these need to be simulated decoupled so that their individual force feedback does not affect each other. In

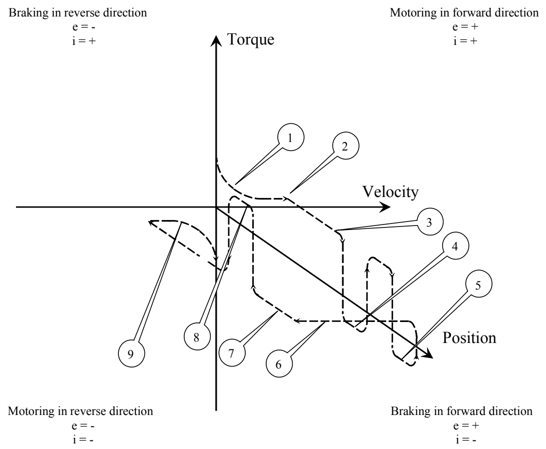

Fig. 4. Four quadrant operation of a dc motor with a given trajectory in PTCA using friction compensation

previous designs of simulators, like in Angioplasty Simulator, [6], or HUI, separation of actuation for both axes could not be achieved because of use of passive actuation, which does not drive the catheter but only does apply force on the catheter.

Along with this study, a simulator with differential mechanism to decouple rotational and translational axes has been conceptually designed as an ideal system for angioplasty simulation because of its potential to comfortably separate primary powered axes. On this design, the main idea is emphasized as preventing crosstalk.

The conceptual design conceives of two main ring gears, one transmission gear, rollers and two DC motors. If we number the components,

I. Upper roller system
II. Lower roller system
III. Rotational ring gear
IV. Transmission gear
V. Translational ring gear

In actuation, DC motors would be used because of their ease of controllability. Two circular main ring gears, which have tooth on inner surfaces, are used to transmit the power to the rollers which holds the catheter from upper and lower sides. These gears are coaxial. Additionally DC motors share the same axis. Two pairs of belt system with two rollers in each lying on the horizontal axis are located inside the ring gears. Ring gears receive their power from DC motors by a string mechanism which circles around the outer surface of ring gears and on the pulley on DC motor axes. The stability and connectivity of the string on the ring gear can be guaranteed and enforced by another set of small

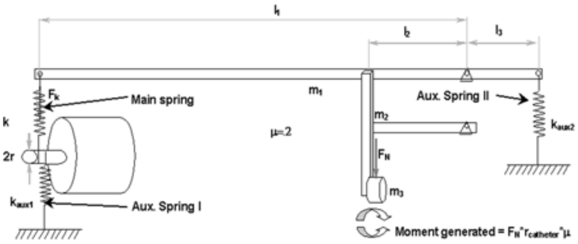

Fig. 5. Sketch of the force actuation mechanism for torsional component

rollers, which pushes down the strings on top of the ring gears' outer surface. Power to the lower roller system is transmitted from the translational ring gear through a small gear, which is rotated 90 degrees with respect to ring gear axis. This transmission gear rotates on the same shaft as lower roller system. Upper roller system does not get powered directly. However, when a tool is driven between two roller systems, because of the spring-loaded upper roller system, the tool is pushed down to the lower roller system.

To avoid crosstalk between force channels, rigid contact between aforementioned main parts are specially arranged. DC motors are rigidly connected to the ground. V is separated from other parts and driven by one of the DC motors. I-II-III-IV are rigidly connected to each other and rotate together.

When translational axis powered alone, upper and lower roller systems rotate and drive the tool. When rotational axis is driven, all four components rotate and transmission gear freely floats on the inner surface of the axial ring gear. However, if not compensated electronically, this motion on the translational gear creates axial motion also. Thus, the axial motion caused by rotational motion should be compensated. Motion details can be summarized as,

- If I-II-III-IV rotates CCW, V should be rotated CCW to compensate the effect of this rotation on translational system.
- If I-II-III-IV rotates CW, V should be rotated CW accordingly to compensate the effect of this rotation on translational system.
- CW rotation of V initiates forward axial motion.
- CCW rotation of V initiates backward axial motion.

With this electromechanical configuration, decoupled translational and rotational axes can be obtained.

IV. CONCLUSION

Regarding the data given by AHA about the occurrence rates of both CABG and PTCA, it is obvious that there exists a growing need for the training of new interventionists in the angioplasty procedure. However, since general training of interventionists happens on live patients, this may put patients at risk. Thus, a simulation environment would be an ideal device for this purpose.

As a first step, design parameters and their effect on haptic perception has been discussed. Inertia, friction and backlash are common concerns affecting transparency of the force reflection. Additionally realizable force resolution should match human hand dexterity for force discrimination. Meeting these requirements, HUI with passive actuation of tools has been proposed. The device has enough force range and control bandwidth to meet general simulation requirements. However, since passive actuation does not provide elimination of remaining imperfections in the system, like friction and inertia, new approach is proposed resulting in a decoupled two axes system and elimination of unwanted forces in force reflections employing differential mechanical drive system for one-tool actuation for simulation. This design will be the basis of further complete angioplasty simulator design and human haptic perception studies in low level of forces used in angioplasty operation.

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